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Designing Embroidered Electrodes for Wearable Surface Electromyography*

A. Shafti, R. B. Ribas Manero, A. M. Borg, K. Althoefer, *Member, IEEE* and M. J. Howard, *Member, IEEE*

Abstract— Muscle activity monitoring or Electromyography (EMG) is useful in gait analysis, injury prevention, computer or robot interfaces and assisting patients with communication difficulties. However, EMG is typically invasive or obtrusive, expensive and difficult to use for untrained users. A possible solution is textile-based surface EMG (sEMG) integrated into clothing and used as a wearable device. This is, however, challenging due to (i) uncertainties in the electrical properties of conductive threads used to construct electrodes, (ii) imprecise fabrication technologies (e.g., embroidery, sewing), and (iii) a lack of standardisation in the choice of design variables. This paper, for the first time, provides a design guide for such sensors by performing a thorough examination of the effect of design variables on sEMG quality. Electrical characterisation and sEMG measurements are performed, considering the effects of manufacturing imprecision. Results show that the imprecisions in digital embroidery lead to a trade-off between low electrode resistance and high consistency. An optimum set of variables for this trade-off is identified and tested with sEMG during a variable force isometric grip exercise with $n=6$ participants and compared with traditional gel-based electrodes. Results show that thread-based electrodes provide a similar level of sensitivity to force variation as gel-based electrodes with about 90% correlation with expected linear behaviour.

I. INTRODUCTION

The human body is a large and complex system formed of different organs working together through different mechanisms in the mechanical and electrical domains. These are all controlled through electrical signals originating in the brain and passed on through the spinal cord. Over the years, medical scientists have found ways to monitor some of these signals and make sense of them, leading to a better understanding of the human body. This is practiced every day in hospitals all over the world from heart-rate monitors and oximeters, to brain activity monitoring through functional magnetic resonance imaging (fMRI) and electroencephalography (EEG). These devices provide real-time information about the patient status to the clinician, however, they do so through obtrusive means, limiting them to stationary scenarios such as the hospital room, i.e., they cannot be used continuously or outside of a controlled low activity environment. If a biological signal acquisition device is to be worn for long term use, it needs to be unobtrusive, lightweight and generally not cause disturbance to the user.

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A. Shafti, R. B. Ribas Manero, A. M. Borg, K. Althoefer and M. J. Howard are with the Centre for Robotics Research at King's College London, WC2R 2LS, London, UK – ali.shafti, roger_bernard.ribas_manero, amanda.borg, k.althoefer, matthew.j.howard@kcl.ac.uk.

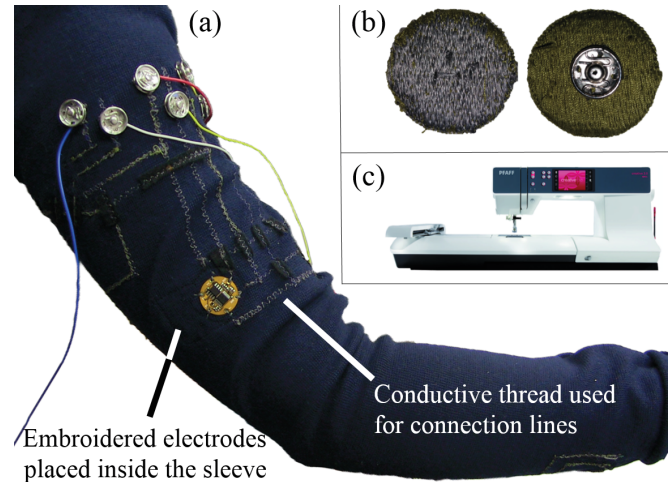


Figure 1 - Embroidered electrodes and their use in wearable sEMG. (a) Sleeve equipped with textile based sEMG, (b) Embroidered electrodes used for the sleeve, (c) Pfaff Creative 3.0 digital embroidery machine used to manufacture the electrodes – machine photo courtesy of PFAFF.

Medical researchers could use such devices to acquire massive amounts of data for better understanding of the human body and consumers could use them to improve their lifestyles in terms of health and wellbeing. There are commercial wearable devices available today, measuring the number of steps, heart rate and sleep activity over extended periods. However, wearable monitoring of electrophysiological data, e.g., brain or muscle activity, is still undergoing research.

Recording muscle activity requires the use of surface electromyography (sEMG) whereby pairs of electrodes are applied to each muscle of interest within a muscle group, and a single ground electrode placed on an unrelated, preferably muscle-free, part of the skin. Proper placement of electrodes is challenging for an untrained user, as the location and spacing between electrodes affects the resulting sEMG signal dramatically. However, if the sEMG sensors are integrated into a wearable platform, such as clothing and textile, these difficulties can be reduced (refer to Fig.1). The user will only have to wear the textile as they normally would, and the sensors will already be placed in the correct positions. The need for reduction of line noise (50Hz/60Hz noise resulting from the power lines surrounding us) and motion artefacts define the challenges in designing appropriate electrodes, electronics and signal processing for such systems.

This paper reports on a step-by-step approach to realising a textile-based sEMG system. Different design variables concerning the fabrication of electrodes through embroidery of conductive thread are examined, and the manner in which they affect the electrical characteristics of the electrodes assessed. The behaviour of the embroidered electrodes during actual sEMG acquisition is evaluated, in comparison with typical gel-based electrodes used in medical applications. The

latter is assessed, through an experiment recording the isometric forces applied in a grip exercise, as well as the muscle activity from the forearm muscles which control the grip. In line with expectations, it is seen that measuring sEMG with embroidered electrodes encounters higher noise levels compared to gel electrodes, however, the variation in force is still distinguishable suggesting the feasibility of using embroidered electrodes in applications such as effort and gait analysis while providing a higher level of comfort and ease of use due to integration into textile and clothing. This paper therefore presents, for the first time, a study to characterise the electrical properties and manufacturability of conductive textile based sEMG electrodes.

II. MOTIVATION AND RELATED WORK

There are a number of benefits to be gained in developing a wearable, textile-embedded sEMG acquisition system. These include (i) continuous, remote monitoring of muscle activity, (ii) ease of use with regards to proper placement of electrodes, (iii) comfort in wearing and unobtrusiveness. For example, in the case of patients with dementia or degenerative illnesses where communication difficulties are present, a wearable platform could be used to ensure the patient's safety. For such patients, wearable sEMG sensors could be used to assess balance through antagonistic muscles. Robot interfacing and prosthetics are also growing areas for use of sEMG. In these cases, proper placement of electrodes as well as long term, comfortable integration of the sensors are of importance which can both be addressed with wearable sEMG.

Many efforts have been made over the last decade on wearable monitoring of muscle activity. In [1] a miniaturised sEMG system is presented and used for wearable purposes, however, the device itself is not wearable nor can it be integrated into wearable devices or clothing. In [2], the same system is developed further but still consists of a bulky band on the muscle and circuitry placed on the user's jacket using tape. In [3], a wearable device for facial expression detection is presented. The wearable interface is placed around the back of the head touching the skin on the cheeks. The paper reports good results on detection of different expressions, however, targets only this specific application and is too obtrusive to be worn continuously in the long term. A more focused effort on creating wearable sEMG systems is presented in [4]. The authors have developed a system with amplification and transmission capabilities in a small size. However, the boards are still bulky and they are not integrated directly into clothing or a small wearable device, but are rather mounted individually on a large band around the forearm. In [5], a multisensory system including a sEMG sensor is created in the form of a wristband, similar to a watch. The latter can be used in long term applications, however, it is not for applications involving integration with fabric and clothing and it is also limited to the specific use on the wrist. Similar work using the wrist watch format is presented in [6] but is dependent on large recording devices. The system developed in [7] is perhaps the one with aims most similar to that reported in this paper. The sEMG sensor is integrated into fabric as a small circuit, using active electrodes and also relies on conductive thread for their circuit connections. However, they rely on capacitive detection of signals as described in [8], [9]. The capacitive

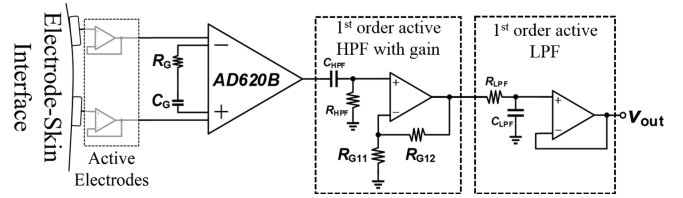


Figure 2 – sEMG acquisition circuit diagram.

approach is useful for this sort of application, as it doesn't rely on contact with the skin, however, this raises some concerns into the integrity of the signals picked up, and introduces new variables into the design approach. In addition, as sEMG acquisition and analysis methods assume contact between the electrode and the skin, it is not clear if the contactless approach can be used with the previously validated analysis techniques and whether results obtained from these sensors will be as reliable. The present study differs by virtue of endeavouring to detect sEMG signals using direct contact with conductive thread based electrodes. This is beneficial as, if achieved, it limits the sources of noise and inaccuracies to those already identified for typical sEMG applications and does not introduce new challenges as is the case with the contactless approach. Direct contact with the skin allows relying on signal processing methods established in previous literature based on similar contact with gel or dry electrodes.

III. SURFACE ELECTROMYOGRAPHY ACQUISITION

A. Skin-electrode Interface Design

There are many approaches to making a sEMG acquisition system, all of which follow the same structure and concept, but with minor changes in terms of inclusion or exclusion of certain optional stages. One of the major factors in the design of such systems, is the ability to eliminate unwanted sources of noise: The design and electrical properties of the electrodes, and the electrode-skin interface are crucial in determining this.

Typically, sEMG systems rely on differential amplification of signals, with *common-mode noise rejection* to minimise noise, where a *common-mode rejection ratio* (CMRR) of 100-120 dB is recommended [8]. In the case of sEMG, this means that the connection from the two amplifier input pins all the way to the signals under the skin need to be matched in terms of electrical impedance. Matching these signal paths, however, is not an easy task as skin impedance is unpredictable and any movement artefacts can affect it. Even the most stable electrode connections to the skin will move as the muscle is contracted and moves under the skin.

The types of electrodes currently in use for sEMG can be categorised as (i) wet and (ii) dry electrodes. In general, silver electrodes with silver chloride electrolyte gel (referred to as Ag-AgCl) are recommended for best sEMG results [8]. Skin preparation also plays a role in the accuracy of the amplification and it is therefore recommended that the skin be cleaned and, in some cases, even shaved before signal acquisition [10], [11]. Note that, in a general purpose wearable acquisition system, intended for use by non-expert users, it cannot be assumed that these skin preparations are performed, adding further difficulty to matching the paths and acquiring a sufficiently noise-free signal.

B. Electromyography on a Textile Substrate

Moving into a textile substrate, for general purpose use presents a number of engineering challenges. The sensor circuit must be simplified to ensure it is small, lightweight and cost-effective, and easily integrated and replaced if necessary. If the system is to be properly integrated, compromises must be made on some of the shielding and noise cancellation techniques common in modern electronics. Furthermore, issues such as the flexibility and stretching of the fabric may affect the resistance of the thread which can cause considerable voltage drops in longer connections and unwanted connections due to fabric bending or fluff from the conductive threads add to uncertainties during circuit analysis and debugging.

As discussed in §II-A, the skin-electrode interface is the most critical block in the systems as it is the largest source of noise [12]. Matching the impedance in this part is challenging, as this is influenced by (i) mismatches between the two areas of skin-electrode contact, (ii) the electrodes are dry, without special skin preparation, (iii) issues such as the flexibility and stretching of the fabric may affect the resistance of the thread. These issues have not been systematically examined in the literature.

The conductive thread used for the electrodes here has a resistance per length specification considerably higher than conventional copper wires ($91.8\Omega/\text{m}$ for the thread compared to $0.0098\Omega/\text{m}$ for copper wires based on relevant datasheets). Thus, the length of the thread that is used in connections plays a larger role in defining the skin-electrode interface impedance, relative to standard electrodes.

Thus the design of conductive thread based *electrodes* for textile sEMG is the main motivation for this study as it is crucial in the functional performance of the sensors.

IV. METHODS AND MATERIALS

The electrodes tested in this paper are made of stainless steel conductive thread (grey colour in Fig. 1(b)), sewn into the fabric with regular thread (non-conductive thread, green colour in Fig. 1(b)) to hold it in place. A Pfaff Creative 3.0 programmable sewing machine is used to make the electrodes. The steel thread is wound onto a bobbin and placed in the bobbin case, and the regular thread is kept on the spool and placed in the spool case. Before the fabrication process starts, the fabric is placed in an embroidery hoop. Hooping the fabric along with a stabiliser keeps an adequate tension required by the embroidery process. During this process, the regular thread is passed through the fabric from one side as it pulls the conductive thread towards the other side of the fabric following the specified design.

The electrodes were designed using the 6D Embroidery System software provided by the sewing machine manufacturer. The tension setting, which defines how tight the thread is sewn into fabric, is set at the maximum possible without breaking the conductive thread ($T=5$). Higher values might not break the thread, but would lead to it going through the fabric in error. Lower tension values were not used to avoid looseness in the conductive thread which would lead to inconsistencies and potential misconnections. A stable contact between different lines of conductive thread is needed for them to be paralleled and reduce overall resistance. The same conductive thread is also used to sew 13mm diameter studs (snap fasteners) to the centre of each

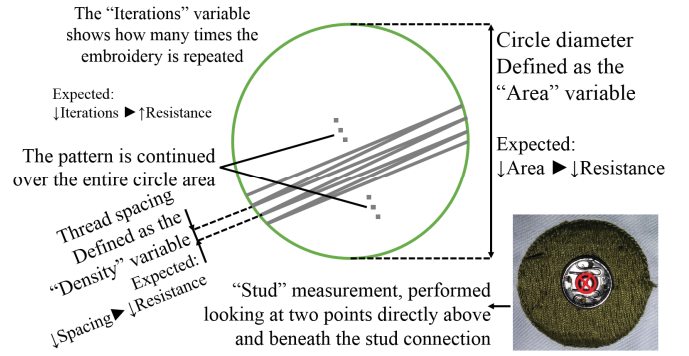


Figure 3 - The inspected design variables shown for clearer presentation. The figure represents a circular electrode during embroidery.

electrode for easy connection to the rest of the circuit (see Fig. 1(b)).

Fig. 2 shows a diagram of the acquisition circuit used for this study. Signals are picked up from the skin using the electrodes. An active electrode system is used where a unity gain buffer is placed on top of the electrodes to assist with impedance matching between the two signal paths. Wire connections take the output of the unity gain buffers to the inputs of an instrumentation amplifier (Analog Devices AD620B). A capacitor is added in series to the gain programming resistor of the instrumentation amplifier (IA) to enforce a frequency varying gain, high pass filtering the sEMG signal as it is amplified. The IA gain is set at 28.5 V/V. The output of the IA is followed by a 1st order active high pass filter (HPF) at 20 Hz to mitigate any remaining low frequency noise such as motion artefacts as well as any potential DC offset from the IA. The active HPF also applies a passband gain of 19 V/V. The HPF is followed by a 1st order active low pass filter (LPF) at 410 Hz as an anti-aliasing filter. The output of this filter is ready to be sampled and converted to digital data. Thus the overall circuit gain is 541.5 V/V. In order to ensure that the new sensors follow validated and established standards in sEMG acquisition, the recommendations of SENIAM (“Surface ElectroMyoGraphy for the Non-Invasive Assessment of Muscles”), a European Union project focused on setting standards for sEMG sensor design, placement and data analysis [13], were implemented. Publications originating from the company Delsys, which provides commercial EMG solutions, were also used to consider factors such as frequency ranges, inter-electrode spacing, electrode placement, reference selection and electrode design during the design and application of the sensor, and the overall acquisition system [14-17].

V. EXPERIMENTS

In this section, the effect of different design parameters on the electrical characteristics of the resulting electrode are investigated. This is followed by sEMG acquisition tests to examine the identified design in a real application.

A. Electrical Property Test

The aim of this experiment, is to identify and characterise the effect of different design parameters on the electrical (i.e., impedance) properties of the embroidered electrodes.

Surface electrodes in general are characterised by physical dimension, shape, technology and constituent materials [13], [18]. When using the sewing machine, aside from the shape and dimension of the electrodes, alternation between different fill densities (i.e., distance between two consecutive thread lines inside the pattern) and different number of iterations¹ is possible.

The three main variables considered are thus (i) the area, (ii) the number of iterations and (iii) the density of the fill in pattern². Fig. 3 shows these design variables and their expected effect on the overall electrode resistance.

Based on the resistance equation:

$$R = \rho (L/A) \quad (1)$$

Where R is resistance in Ohms, ρ is resistivity, L is the length and A is the cross sectional area of the wire, it is expected that shorter lengths of thread and parallel connections between threads (due to paralleling of resistances) will result in lower overall resistance and thus lower impedance for the interface. This translates into smaller electrode area, higher number of iterations and higher density as the predicted optimal electrode design. While a higher number of iterations is expected to result in a stiffer electrode, it does not lead to issues with conforming of the electrode to the shape of the body due to the small area of the design.

A set of electrodes varying each of these parameters were fabricated using the Pfaff sewing machine as described in §IV. With regards to (i) area, results are reported for electrodes of diameter 1cm, 2cm and 3cm, (ii) iterations, those for electrodes fabricated with 1 (no repeats), 2 and 3 iterations are reported, and (iii) density of the fill-in pattern, thread spacing of 5mm, 4mm, 3mm, 2mm and *full* (no spacing) are reported.

In order to see how each parameter affects the overall electrical behaviour of the electrode, each parameter is varied independently, and results are compared against a baseline electrode design, representing the mean values for each variable, i.e., 2cm diameter (medium area), 2 iterations and 3mm density spacing.

To characterise the impedance properties of the electrodes, the resistance in the path from the stud, to the centre point on the electrode surface (refer to Fig. 3) is measured using a digital ohmmeter. Each measurement is taken N=20 times for each electrode, followed by ANOVA to determine whether the mean resistance value varies significantly or not according to each parameter varied.

The resistance measurements are reported in Table 1 across different variations in the design variables. Electrodes are labelled using the code: (*area, iterations, density*), e.g., the baseline electrode -- which has a 2cm diameter, 2 iterations and a density spacing of 3mm -- is marked as (2,2,3).

Looking at the effect of area variation, a decrease in area (line 2) does not result in a statistically significant variation in resistance, but an increase in area (line 3) results in a significant decrease in resistance. In terms of iterations, a

Table 1 – MEAN, STANDARD DEVIATION AND P-VALUES OF THE RESISTANCE FOR DIFFERENT ELECTRODES.

#	Electrode	Mean	SD	P-Value	
1	(2,2,3)	0.30	0.05	-	Baseline
2	(1,2,3)	0.27	0.05	6.54×10^{-2}	Area Variation
3	(3,2,3)	0.22	0.04	1.01×10^{-5}	
4	(2,1,3)	0.33	0.04	5.48×10^{-2}	Iteration Variation
5	(2,3,3)	0.34	0.07	2.36×10^{-2}	
6	(2,2,5)	0.19	0.06	2.71×10^{-7}	Density Variation
7	(2,2,4)	0.35	0.08	1.99×10^{-2}	
8	(2,2,2)	0.19	0.06	2.71×10^{-7}	
9	(2,2,full)	0.19	0.03	6.52×10^{-9}	

decrease (line 4) makes no significant difference but an increase (line 5) results in a significant increase in resistance. During density variation, a decrease from the baseline (line 7) results in an increased resistance and an increase (line 8) will result in a decreased resistance. Further density increase (line 6) as well will result in decreased resistance. Further density reduction also results in decreased resistance (line 9).

The results generally follow the theoretically expected trend, but there are also some unexpected results. These anomalies can, however, be explained if the effects of the embroidery machine manufacturing inaccuracies are taken into account. The embroidery machine is not entirely accurate at low dimensions and spacing. Some manufacturing errors are therefore unavoidable, resulting in the conductive thread not paralleling with itself as is expected and leading to longer series connections and higher overall resistance values than expected.

In the case of area variation, no change in resistance is expected as it is measured between two points directly beneath each other (refer to Fig. 3). However line 3 in Table 1 shows a significant change in resistance. Considering the small size of the design and the anomalies resulting from it, it can be deduced that an increase in area will result in a more relaxed design for the embroidery machine due to the increased dimension. This means there are less errors in the manufactured electrode and therefore a lower resistance is observed. For iteration variation, the resistance is expected to decrease with increased number of iterations, but this is not observed in the results (Table 1, line 5). Adding another iteration of the same pattern within the small dimension of the electrodes results in a crowded design that can lead the machine into errors. These errors can explain the increased resistance. During density variation, lines 7-9 of Table 1 present the expected result of decrease in resistance with increasing density. However, a decrease in density is expected to increase the resistance, which is not the case (Table 1, line 6). As the design gets denser, manufacturing becomes more difficult and errors can be made by the embroidery machine. This can explain why a lower resistance is witnessed when the density is reduced as it results in a less complex design.

The above errors mean that in low dimensions and crowded designs, the results expected by the theoretical analysis do not match. There is a trade-off between achieving lower resistance and keeping manufacturing errors to a minimum that can also lead to inconsistencies. While the digital embroidery machine used for this work is high-end, there is the possibility that more advanced machines will result in less such errors. However, there will always be an inconsistency in results because of manufacturing errors,

¹ Iterations represent the number of times the embroidery is repeated on a single electrode.

² Preliminary experiments showed similar behaviour between different electrode shapes (circle, square, and rectangle), and so this factor was discarded from the study. In the following, results are reported for circular shaped electrodes.

however small they might be. Thus, a trade-off will be faced even if other machines are used.

Based on these experiments, the designs with the best results (lowest mean impedance) are those with 2cm diameter (medium area), 2 iterations (medium number of iterations) and 2mm, 5mm or *full* spacing grid pattern (lines 6, 8 and 9, respectively). In the following section, the (2,2,2) design is evaluated for sEMG acquisition in a simple isometric force task.

B. sEMG tests

Having characterised the electrical properties of the electrodes according to the design variables (ref. §IV-A), in this section, their performance for sEMG acquisition is examined in comparison to conventional medical gel-based electrodes. The aim is to evaluate the use of the conductive thread-based electrodes in terms of distinguishing between different levels of force applied by the muscles.

For this, the experimental procedure is as follows. Pairs of electrodes are placed on the participant's forearm muscles, one pair on the flexor muscle group and another pair on the extensor muscles. A dynamometer (Camry 90kg Digital Hand Dynamometer) is fixed to a table as an exercise tool. The participant rests their arm on the table, and the arm is fixed in place using strapping in a position where they can grip the dynamometer with comfort. Participants are then asked to apply grip forces to the dynamometer. As the arm is fixed to the table, the exercise can be considered isometric and therefore a linear relationship between sEMG and force levels is expected [19]. The dynamometer allows the monitoring of the applied force level. The active electrode circuit used to pick up the electrode signals (Fig. 2) is housed inside a box with handles that allows it to be fixed onto the participant's arm similar to an armband. The box also houses stud connectors that can connect to both the conductive thread electrodes and the gel ones (Covidien Kendall Arbo H124SG), see Fig. 4. In this manner, stability and fair comparison between exercises is ensured. While the electrodes were fixed to the arm during the experiments, in an actual wearable application, they would be sewn into tight fit fabric, such as sportswear, which would ensure the electrodes stay in place, at the right location, without discomfort. The outputs of the circuit are connected to a Bitalino microcontroller system which samples the signal at 1 kHz and sends the data wirelessly through Bluetooth to a nearby computer. The signal is recorded in the computer and further analysed using MATLAB. Fig. 4 shows the experiment setup.

Each participant applies a set of 6 force values on the dynamometer while their muscle signals are recorded. The participants are asked to look at the force value displayed on the dynamometer and attempt to hold the 6 specific force values, each for 10 seconds, with 10 second rests in between. The force values were³ 5kgf, 7.5kgf, 10kgf, 12.5kgf, 15kgf and 17.5kgf. This exercise is repeated 5 times with the embroidered electrodes and 5 times with the gel-based electrodes – 10 exercises overall – exercises with 5 minute rests in between exercises to avoid fatigue. The results reported here are those obtained from n=6 participants (3

³ Kgf (Kilogram-Force) defined here as equivalent force for a certain weight value, i.e. 1kgf = 9.8N. Therefore the force values in Newton are 49N, 73.5N, 98N, 122.5N, 148N and 171.5N respectively.

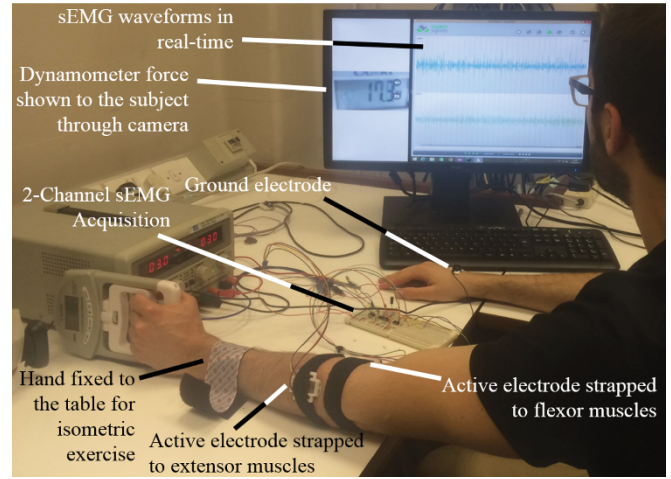


Figure 4 - Experiment setup. The active electrodes are held by hook and loop strap and are touching the participant's skin through embroidered electrodes.

male, 3 female). Participants vary in age (23 ± 6 years old), gender, race, dominant hand and hours of exercise per week. Ethical approval was obtained prior to the tests (reference number BDM/13/14-123).

From the raw data, basic pre-processing is applied to get a clear signal. First, the signal is divided into equally sized sections corresponding to each single force application and the mean value of each is removed (i.e., the DC set to 0). The resulting signal is then scaled and attenuated to show the actual sEMG value on the skin interface before the circuit gain is applied. This is then high pass filtered using a 4th order Butterworth filter at 20Hz. The filtered signal is rectified by obtaining its absolute value and smoothed by computing the root mean square (rms) of values in a sliding window to obtain a linear envelope. The window size used here is 200ms. Finally, the average value for the linear envelope during the force application is calculated and used as a measure of the sEMG level for that particular force. In this manner, each single force value has a corresponding single sEMG value associated to it⁴.

Fig. 5 shows the sEMG versus applied force data for one of the participants' flexor and extensor muscle groups for the thread (top row) and gel (bottom row) electrodes, and a trend line. Different markers represent different trials. As can be seen, the gradient of the thread-based and gel-based trend lines are quite close to each other. The trend line for the conductive thread based electrodes has a 9.4% higher

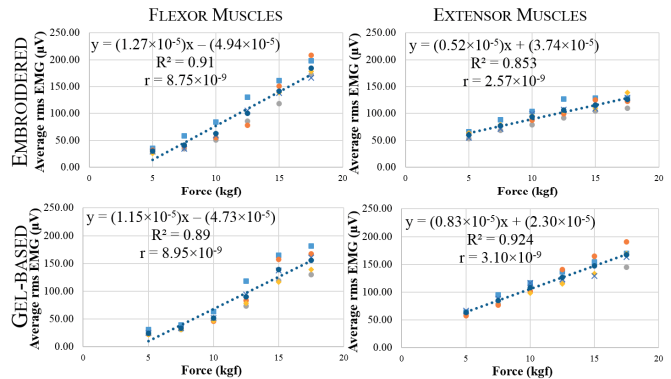


Figure 5 - EMG values versus force levels for different electrodes tested with one of the six participants' flexor and extensor muscle groups.

⁴ Data is available on request.

Table 1 - VALUES CORRESPONDING TO LINEAR REGRESSION FOR THE EMG DATA ACQUIRED FROM ALL SIX PARTICIPANTS.

Type	Flexor Muscles			Extensor Muscles		
	Gradient	R ²	r	Gradient	R ²	r
Emb.	6.76×10 ⁻⁶	0.519	1.39×10 ⁻⁷	5.34×10 ⁻⁶	0.431	1.24×10 ⁻⁷
Gel	5.54×10 ⁻⁶	0.474	1.11×10 ⁻⁷	7.05×10 ⁻⁶	0.390	2.55×10 ⁻⁹

gradient when compared to the gel electrodes in the flexor group, showing more sensitivity to force variation. This is 37.1% lower in the extensor group, where gel electrodes have shown a higher sensitivity to force variation when compared to the thread electrodes. R-squared is calculated to verify the correlation between the EMG and the measured force, showing 91% correlation for the thread-based electrodes and 89% for the gel-based in the flexor group. Sum of squared residuals (r) is also given in Fig. 5. This is generally less for the embroidered electrodes, showing better fit with the estimated trend line. These results show that the conductive-thread based electrodes have been effective in providing similar results to those of gel based electrodes.

Table 2 shows the results of the experiment for all 6 participants. As indicated in table 2, the gradient of the trend lines are similar; 17.9% higher for the thread electrodes in the flexor group compared to the gel electrodes. R-squared results show 51.9% correlation for the conductive thread-based electrodes and 47.4% for the gel-based in the flexor muscle group. Similar results can be seen for the extensor group however with lower correlation values and in this case a 24.2% higher gradient value for the gel-based trend line compared to the thread based. Overall, results show that the different force levels can still be distinguished using the conductive thread-based electrodes prepared as part of this study.

Similar behaviour was expected for the flexor and extensor muscle groups. However, in the case of the extensor muscles, the thread electrodes are showing worse performance when compared to gel electrodes. This might be due to the extensor side of the forearm being hairier than the flexor side, resulting in more noise for the dry thread electrodes when compared to gel electrodes that adhere to the skin even in the presence of hair.

In an extended version of this paper⁵, the results of the study have been used to create jogging leggings capable of continuous sEMG measurements to study the behaviour of leg muscles during running in outdoor scenarios. Using sportswear allowed for a tight yet comfortable fit, ensuring that the electrodes are at the correct location and that limited skin travel occurs, minimising motion artefacts on the signal. The outdoor scenarios also allowed the testing of the embroidered electrodes in dynamic motions, showing results consistent with expectations. Thus, the method described in this study can readily be used to create wearable sEMG measurement garments for different muscles in the body.

VI. CONCLUSION

This paper provides, for the first time, a thorough analysis of electrical characteristics and behaviour of embroidered sEMG systems. Through this process, anomalies due to manufacturing inaccuracies of the digital embroidery machine at low dimensions were identified, that lead to

results not predicted by theoretical analysis. These need to be considered during the design of embroidered electrodes. The embroidered sEMG system was tested with force-EMG relationship experiments proving the feasibility of its use in acquiring measurements of muscle activity.

These results enable future research in the area that can lead to a wearable garment with integrated sEMG sensors. Such a garment can serve as a low-cost and easy to use interface for its wearer to interact with computers or robots, especially for people with disabilities. For robotic prosthetics in particular, textile sEMG would lead to easy integration with daily life without concerns on sensor placement accuracy.

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⁵ A preprint is available on request.